Microcomputer Control Of An Above Elbow Prosthesis

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ABSTRACT

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Simultaneous control of several degrees of freedom of multifunctional prostheses that can be subconsciously operated has been a major problem in prostheses design. To date no system has been fully successfull in achieving this goal.

This thesis describes a control scheme for prostheses, that, after a short training period, is expected to reduce considerably the requirement for amputee concentration in operating his multifunctional prosthesis in a coordinated manner whilst providing him with sufficient functionality.

The proposed control scheme is a modified form of extended physiological proprioception. Programmable fixed linkages are used to achieve coordinated control of several degrees of freedom of the prosthesis.

A microcomputer controlled above-elbow prosthesis using the above control method has been designed and a bench model has been built. The controlling input to the prosthesis is the shoulder position. The controlled functions are: hand prehension, wrist rotation, and elbow flexion/extension.

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1. INTRODUCTION

Replacement of lost limbs and their functions by artificial equivalents has been a major challenge for rehabilitation scientists and engineers. Although, to date, no system has been successful in matching, or even coming close to, the mechanical performance of the natural arm, considerable progress has been made in enhancing the capabilities of the amputee.

Early prostheses used body power to control the functions of the artificial arm. They were developed to a high degree in America during the early 1950's. Basically they use either arm flexion (glenohumeral flexion) or shoulder shrug (scapular or biscapular abduction) to operate a Bowden cable that activates the artificial limb.

A body powered prosthesis for the above elbow amputee is usually provided with a flexible elbow and a prehension hook mounted on a friction turntable to enable passive wrist rotation. Both the elbow and the hook are activated by rounding the shoulders to tighten a strap, thus pulling the Bowden cable. A shoulder activated mechanical switch locks the elbow in place allowing further shoulder action to perform hook opening. Elbow extension uses gravity and the hook is spring loaded in the closed position.

A typical body powered above-elbow prosthesis is shown in figure 1.1. A body powered prosthesis often provides good function for the patient. It has the advantages of being simple, rugged, relatively inexpensive and easily repairable.

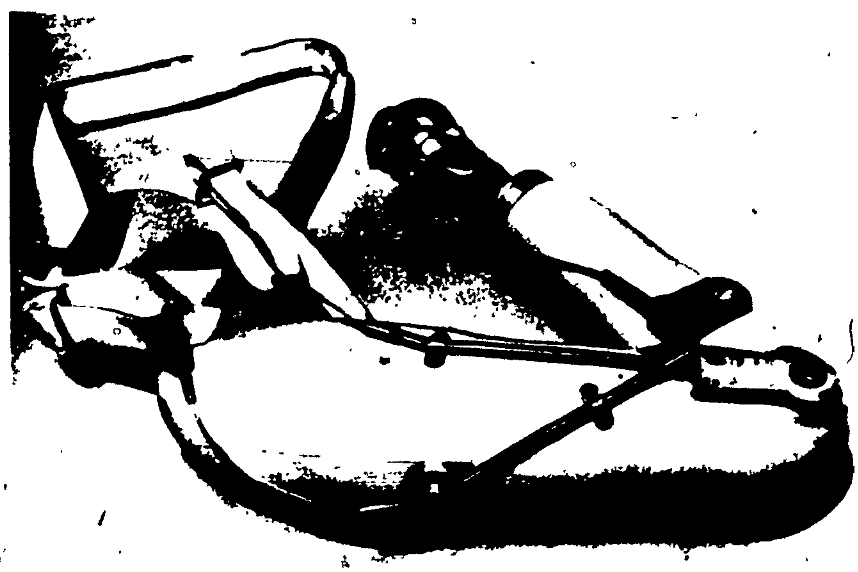

Fig. 1.1: A typical body powered above elbow prosthesis.

The major disadvantage of a body powered prosthesis is its cumbersomeness to wear because of the harnesses. The need for excessive harnessing may also be uncomfortable and restrictive in terms of the envelope of space with respect to the body in which the prosthesis may be operated. Gross motions are required to activate the arm. The pinch forces in the hand and the lifting forces on the elbow are not as large as may be desired. Restrictions on coordinated motion, such as the impossibility of simultaneous operation of the elbow and the hook, are other drawbacks of the body-powered systems.

At high levels of amputation, such as shoulder disarticulation, bilateral disarticulation or congenital absence of both upper extremities, body powered positioning of artificial arms and opening/closing of the terminal device is not possible [18]. Hence for such cases external power becomes indispensable.

In an externally powered prosthesis, the joints of the artificial arm are driven by externally powered actuators. The actuators that have been considered are either pneumatic or electrical.

The first externally powered prostheses were pneumatic. In 1957 Marquardt and Hafner first fitted a child with bilateral amelia of the upper limbs with a pneumatic prosthesis [18]. During the 1960's, pneumatic arm prostheses controlled by body-activated valves were being used rather extensively in Europe by limb deficient children [4]. In America, however, the trend was towards electric power, mainly due to the absence of a good distribution system for compressed-gas replacement cylinders.

One disadvantage of electric actuators is their heavier weight compared to compressed-gas actuators. However, new design ideas incorporating small electric motors coupled with complex gear systems promise a considerable reduction in actuator weight. Also, the relative ease of recharging batteries as opposed to changing or refilling the compressed gas cylinders, makes electric prostheses more attractive.

A prosthesis, essentially, is a machine attached to the man. The man has to send commands to the prosthesis to perform the desired functions. In an externally powered prosthesis, the control signals are usually of biomechanical or bioelectric origin [4]. Among the biomechanical sources that have been investigated, the most acceptable one seems to be the motion of a part of the body. If, for example, the shoulder motion or the motion of the clavicles is used to control the prosthesis, the amputee can learn to associate his shoulder or clavicle positions with the functions of the prosthesis.

A bioelectric control source currently under consideration is the myoelectric signal (EMG). In myoelectric control, muscle potentials are used to control the functions of the prosthesis. These signals occur when the brain sends a message to a muscle and the muscle tissues expand or contract to produce the requested motion. In a large number of amputations, it is known that the amputee retains a residual image of the lost limb, often complete with sensations. When the amputee tries to move this phantom limb, EMG impulses occur in the remaining musculature. These impulses form patterns which are distinguishable with regard to intended motion. In myoelectric control, they are detected by means of electrodes inserted into the muscle or via skin electrodes applied to the skin over the muscle.

The major problem in the design of multifunctional prosthesis is the simultaneous control of several degrees of freedom. As the degrees-of-freedom, or the number of prosthesis joints that can be independently controlled increases, the complexity in the coordinated control of these joints also increases in a non-linear way. A high degree of user concentration becomes necessary to control the prosthesis.

Various methods to achieve subconscious coordinated motion in several degrees-of-freedom using control signals of biomechanical or bioelectric origin have been suggested. They will be examined in the next section.

2. CONTROLLING CONCEPTS

From a theoretical point of view, the most attractive approach towards coordinated motion is to use the brain's own "computer" to perform the control [20]. Then, the signals from the brain that were used to control the normal arm will be directly employed to coordinate the various degrees of freedom of the artificial arm. EMG is one such signal which is currently in common use.

Earlier myoelectric controlled prostheses looked only for the presence or absence of the EMG signals at the controlling sites [16]. Thus, if the biceps and the triceps muscles were used to control the opening and closing of an elecric hand prosthesis, respectively, then the electrical activity on the biceps muscles was interpreted as a command to open the hand while the electrical activity on the triceps muscles was interpreted as a command to close the hand. If more degrees of freedom on the prosthesis, such as elbow flexion/extension and wrist pronation/supination were required, then more muscle sites would be used to control the additional degrees of freedom.

Recent advances in technology, such as the development of microprocessors, have made the on-line analysis of the EMG patterns possible. The information obtained from this analysis is used to control the various degrees of freedom of the artificial arm.

There are two major approaches to match the various myoelectric patterns to desired limb functions. The first, based on the works of Lawrence [14], Lyman et al [5],[17] or of Jacobson and Mann [12], requires mapping of several electrode sites, at each of which the EMG function is strongly correlated with a single limb function. It employs the average levels or the low-frequency characteristics of the EMG signals and their mapping or distribution over the various electrode locations.

This approach requires many electrode sites, usually of the order of ten or more depending on the number of limb functions to be independently controlled. For example, the system suggested by Lyman and Freedy [5] uses EMG signals from nine sites to control three degrees of freedom of the artificial limb. Due to the number of electrode sites required, this approach is limited to amputees with little nerve and muscle damage to their stump and with relatively long stumps.

A second approach to myoelectric pattern matching has been suggested by Graupe and his associates [7,8,9], which utilizes the signal's statistical dynamics rather than its level. Since the method is concerned with the complete spectrum of the signal at a given site, such that the complete linear information content of the EMG signal is considered (i.e. at all frequencies), it is considerably more efficient in terms of utilizing the information content of the EMG signal, and thus less electrode locations are required [7]. The requirement for less electrode sites makes this approach suitable for amputees with shoulder disarticulation or with severe nerve and muscle damage to their stump. In their system, Graupe and his associates use from one to three muscle sites to control five limb functions.

One major problem with EMG control is that its operation is highly dependent on direct visual feedback. There are basically two reasons for this dependence. The first reason is due to the fact that EMG is exclusively an efferent phenomenon [10]. In other words, if we regard the man/prosthesis system as an interaction between man and machine, EMG is capable of providing communication from man to machine but not vice versa. The second reason for the necessity of visual feedback is that the activity of the muscles is represented only poorly, if at all, at the conscious levels in the central nervous system. Thus an amputee using an EMG controlled prosthesis would be poorly aware of what he was telling the machine to do and hence would have to rely heavily on direct observation of the prosthesis.

This dependence on visual feedback may be acceptable if the controlled joint is not primarily a position function such as the control of hand prehension. However, if the positioning of the artificial arm in space relative to the body is concerned, the lack of proprioceptive information makes the amputee very dependent on visual feedback.

As the degree of freedom of the prosthesis increases, a large amount of user concentration is required. This is very undesirable since the main occupation of the amputee should not be the control of the prosthesis, but rather he should be allowed to perform other intellectual activities such as talking or thinking while he is performing some task with his artificial arm. Thus, the control of the prosthesis has to be subconscious in nature as is the case in the control of the natural arm.

Efforts to provide sensory feedback by means of externally generated stimuli have been considered by various researchers [17]. The use of pressure on the skin, or the application of electric pulses proportional to the signals originating at the prosthesis control site, may be useful to some extent for force feedback. However, it is totally inadequate as a means of supplying position information. The lack of such proprioceptive information, and thus dependence on visual feedback, leads to the rejection of multifunctional prostheses by the majority of amputees [10].

EMG controlled arm prostheses developed to date can be considered open loop systems since no satisfactory position feedback, apart from visual feedback, has been provided to the amputee. Consequently, attempts to achieve subconscious coordinated motion via myoelectric control have not been successful to date. However, other techniques which are less elegant but more promising have been investigated.

One such technique for achieving coordinated motion is to use a system of linked motions so that coordinated motion of several joints is automatically coupled to produce a certain function with only one control signal input [13]. Mechanical, hydraulic or other forms of linkages can be implemented so that a single input signal causes the coordinated activation of several joints. This type of fixed linkage, however, is likely to be constraining to the amputee since he will find it easy to perform a certain task with his prosthesis, while to perform another will be impossible. Carlson[3] has investigated the possibility of generating a pattern of linked motions which is useful for several common tasks.

Another alternative for simultaneous subconscious control of several degrees of freedom is to obtain position feedback from all the artificial joints. A one-to-one correspondance is established between the joint angles of the artificial limb and the natural joints of the body that control those artificial joints. The patient obtains proprioceptive information (or position feedback) from the natural joints, thus eliminating the requirement for visual observation and excessive concentration.

The validity of this approach is demonstrated by the excellent sensory feedback obtained by an amputee using a body powered prosthesis incorporating a Bowden cable. Force, movement, and position information are fed back to the amputee in a very natural and easily comprehensible way. This system works very well because the movement, force and position of the prosthesis is related on a 1:1 basis to the movement, force and position of the Bowden cable. Simpson [21,23] achieves a similar quality of proprioceptive information with his CO_2 powered prosthesis . In designing his prosthesis, Simpson introduced the term "extended physiological proprioception" (e.p.p.) which refers to the establishment of a direct relationship between the movement of a normally functioning joint (the input signal) and the movement of the prosthesis. In his prosthesis, Simpson uses the position of the clavicular joints as inputs, thus extending the natural position feedback (proprioception) of the clavicular joints into the prosthesis.

This concept of extended physiological proprioception is analogous to the position and velocity information obtained through the hand and other body joints involved in sports such as tennis, squash or baseball. The person learns to associate his body joints with the position of the artificial extension (i.e. the raquet) and control becomes subconscious in nature. Similarly a patient can learn to associate the position of the artificial arm with the position of a normally functioning joint of the body (the clavicular joint positions in the case of Simpson's prosthesis).

Simpson also introduced an additional feature termed the "unbeatable servo" to prevent the control input from moving faster or further than the prosthesis can follow. This constraint on the input to move only as fast as the prosthetic joint, helps to preserve the one-to-one relationship between the input and the output which is essential to e.p.p.

Other groups have also used the principle of e.p.p. in the design of electrically powered prostheses [1,25]. Like Simpson, they use clavicular joint positions as input signals. Prosthesis control systems employing e.p.p have had a dramatic effect in the reduction of the training period and in the acceptance of the arms by the amputees [22].

3. DESIGN CRITERIA FOR OUR ARM

After reviewing the various options for the control of multifunctional prostheses, and based on their experience in the field, Gibbons and O'Riain [19,6] decided to set down the following priorities for prosthesis design:

(i) Proprioceptive feedback of force, movement and position should be made available to the amputee.

(ii) An easy-to-learn means of achieving coordinated movement of the prosthesis should be implemented.

Even though, from a theoretical point of view, the most elegant means of achieving these objectives is via myoelectric control, the open loop nature of such control modalities (due to the absence of afferent pathways) makes it impossible to realize a myoelectric controlled multi-functional prosthesis that will meet the first of the above priorities.

At the present time, and probably for some time to come, the only truly satisfactory modality of position proprioception remains to be one which makes use of a limb position. The validity of this is demonstrated by the wide acceptance the body powered prosthesis and externally powered prostheses using e.p.p. has found among the amputees.

Hence Gibbons and O'Riain decided to use the principle of extended physiological proprioception in the electrically powered above elbow prosthesis. However, a major modification to the conventional application of e.p.p. was introduced.

In a conventional e.p.p. system, each prosthesis joint is controlled by one normally functioning joint [4]. If the simultaneous motion of two or more prosthesis joints is required, the amputee has to move two or more of his corresponding joints in a coordinated manner. Even though the amputee receives proprioceptive information regarding the position of each prosthetic joint, their coordinated control requires some concentration.

The modification introduced by Gibbons and O'Riain involved the adoption of a system of linked motions such that one normally functioning body joint motion will result in the coordinated motion of two or more prosthetic joints. This feature would relieve the amputee from the burden of coordinating various prosthetic joints.

The linkages would be pre-determined and be selectable by the amputee. A microprocessor system with stored programmes was to be used for this purpose. The selectability of various linkages overcomes the difficulty imposed by conventional linked motion prostheses, where the amputee is easily able to perform a few tasks with his prosthesis while he finds it impossible to perform others. Using the approach of Gibbons and O'Riain, the amputee will be able to choose the proper linkage for the task he wishes to perform.

The selection of different linkages will be achieved by a simple switching system that will be activated by the other (usually intact) hand. Frequent switching between different linkages will not be necessary since tasks requiring manipulation are often repetitive [6].

For the above-elbow prosthesis, Gibbons and O'Riain decided to use the shoulder joint as the controlling input. The choice of the shoulder position as the input signal was based on the recognition of the fact that many manipulative tasks involve a coordination of the shoulder, elbow, and wrist movements. Thus the use of the shoulder joint, which would be naturally involved in the movement of the normal arm, will help to make the artificial arm movements resemble natural arm movements.

One drawback of the use of the shoulder movement as the controlling input is that the movement of the wrist and elbow becomes impossible without the movement of the shoulder. In the situation where an amputee is greatly involved in activities that require the motion of the wrist and elbow without the movement of the shoulder, other joints such as the clavicular joints will have to be used as the controlling input to the prosthesis.

The motion of the clavicular joints has been successfully used by Simpson in his e.p.p. prosthesis and will be used by us to extend our prosthesis for shoulder disarticulation amputees. However, we will limit our discussions at this point to above elbow prostheses where the shoulder joint can be used as the controlling input.

An under-arm goniometer which measures shoulder flexion and extension will be used to provide the input to the microprocessor. Upon receiving information on the shoulder position, the microprocessor will position the controlled joints according to the selected linkage. A different choice of linkage will result in a different positioning of the artificial arm for the same shoulder position. Thus by selecting different linkages, the amputeee will be able to vary the input-output relationships of the system. Hence he will be able to perform a variety of different activities.

It is expected that the amputee will be able to learn, in a short training period, the various linkages that are available to him. By learning those linkages which are essentially the input output relationships of the prosthesis, he will be able to associate his shoulder joint with the position of the aritifical arm. Hence he will have excellent proprioceptive feedback.

Furthermore, having selected a certain linkage such as bringing objects to the mouth, all he will have to do is to move his shoulder as he would normally do to perform this task. The artificial arm joints will be positioned as defined by the pre-programmed linkage. Hence, simultaneous coordinated motion with proprioceptive feedback will be achieved.

In addition, it was decided to incorporate Simpson's "unbeatable servo" concept in the new prosthesis. The input goniometer would be equipped with a braking mechanism. The braking will limit shoulder movement in accordance with the current output capabilities of the system. Thus the 1:1 relationship between input and output, which is essential for e.p.p. to be effective, will be continuously preserved. The role of this braking mechanism will become less and less important as the amputee grows accustomed to the capabilities of his prosthesis. Once this has been achieved, the control of the prosthesis will become very close to subconscious.

An additional advantage stemming from the use of a microprocessor is that the dynamic characteristics of the actuators can be put under program control. This means that the amputee can regulate the speed of his prosthesis by varying the speed of his shoulder movement. The maximum speed, however, will be determined by the current output capabilities of the prosthesis. An attempt to demand more than the system can handle will be prevented by the braking mechanism.

Furthermore, the microprocessor system allows the programming of different sets of linkages to fit different patients needs. According to the environment in which the amputee wishes to use his prosthesis (such as home and different kinds of work), he may be provided with custom-built pre-programmed linkages.

A prosthesis using the above approach has been designed and a prototype has been built for above-elbow amputees. It will be described in the following chapter.

4. DESCRIPTION OF PROTOTYPE PROSTHESIS

The prosthesis that will be described in this chapter has been designed for above-elbow amputees. A bench model has been built for laboratory test purposes. A prototype to be tested on amputees for clinical evaluation of the prosthesis is currently under development.

The controlling input to the prosthesis under consideration is the shoulder position. It is assumed that the amputee will be able to use his shoulder joint to control the prosthesis. The controlled functions of the prosthesis are: hand prehension, wrist rotation (pronation and supination), and elbow flexion/extension. Elbow and wrist joints are controlled by the modified e.p.p. system described in the previous chapter. Hand prehension is to be controlled by EMG using a system developed at the Rehabilitation Institute of Montreal (RIM) [15]. This assumes that the amputee's biceps and triceps muscles are in reasonably good condition to provide the necessary input signals.

The prosthesis consists of five major components. These are:

1) The hand unit and its controlling electronics
2) The wrist rotation unit

3) The elbow unit

4) Goniometer with a braking mechanism

5) The microcomputer and the supporting electronics

Standard commercially available hand, wrist and elbow units were used to construct the prosthesis. An Intel 8748 microcomputer with the minimum of support electronics provides the control for those units.

In the following sections a brief description of the prosthesis components will be given. Then in the next chapters the controlling software and the support electronics will be described in detail.

4.1 Hand Unit:

The hand unit that is being used is a simple pinch prehension hand developed by the Otto Bock company. It is controlled by EMG signals from biceps and triceps muscles using the system developed at the Rehabilitation Institute of Montreal. This hand unit is routinely being applied at RIM to below-elbow amputees with proven success. There is no sensory feedback incorporated with the unit, and its simple prehension falls far short of the full functionality of a natural hand. Nevertheless, it has found considerable acceptance among amputees. The hand unit along with the controlling electronics is shown in figure 4.1.

One of the attractive features of the RIM control strategy is its flexibility of utilization. It accommodates either myoelectric control or a displacement transducer which could be adapted to some anatomical movement [15]. This versatility will allow us to use some form of biomechanical control for the hand unit in the case where the biceps and triceps muscles on the amputee's stump are too badly damaged to provide the necessary EMG signals for myoelectric control.

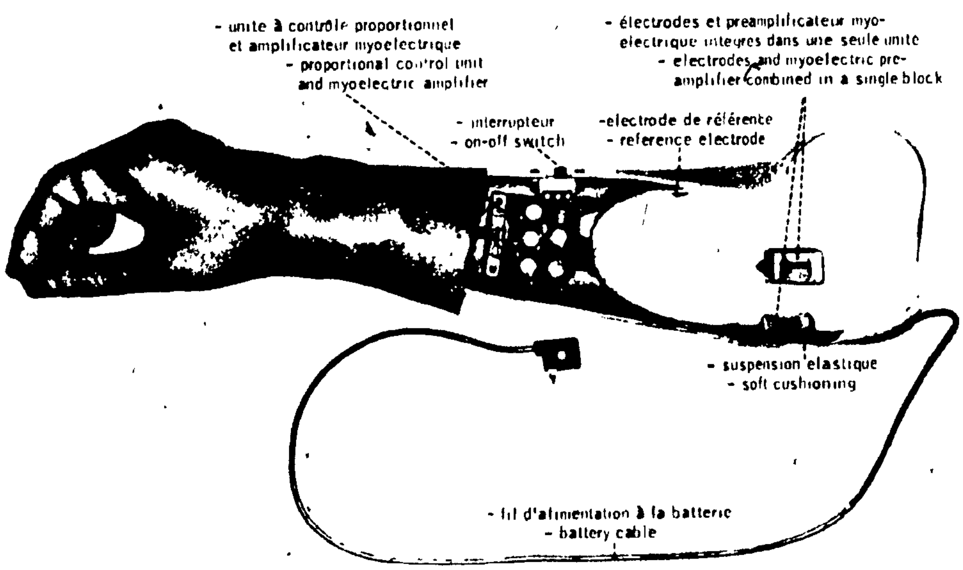

Fig. 4.1: RIM hand unit with it's controlling electronics.

4.2 Wrist Unit:

The wrist rotation unit being used in our prosthesis is also developed by the Otto Bock company (Figure 4.2). It provides simple rotation which has been constrained to movements between pronated (palm down) to supinated (palm up) positions via microcomputer control. The modification that we have made to the wrist unit is the addition of a linear potentiometer to provide feedback of the wrist position to the controlling microcomputer. The microcomputer generates the necessary signals to activate the wrist motor and controls its direction of movement by switching the polarity of the signals applied to the motor.

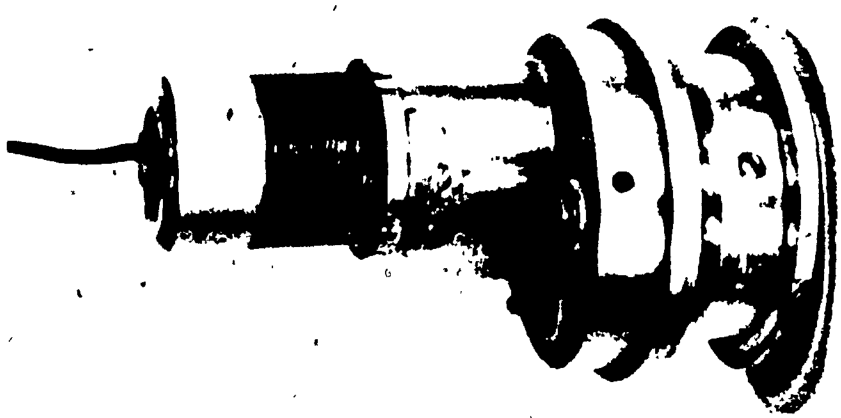

Fig. 4.2: The Otto Bock wrist unit.

The flexion and extension of the wrist is not possible with our current wrist unit. However, this motion can be included at a later stage using friction joints that will provide passive flexion and extension of the wrist. This may be expected to improve the functionality of the prosthesis.

4.3 Elbow Unit:

In our bench model we have used a Variety Village (VV) elbow developed for children to provide the elbow flexion/extension of our prosthesis (Figure 4.3). A feedback potentiometer has been added to this elbow to provide position information on the elbow joint to the microcomputer.

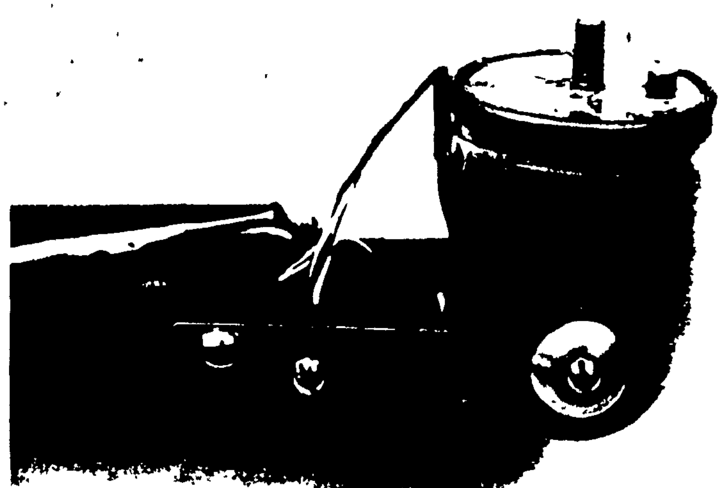

Fig. 4.3: Variety Village elbow for children.

In the construction of the prototype, however, the VV elbow unit is being replaced by the Boston Elbow developed by Liberty Mutual Assurance Company in collaboration with M.I.T., Boston (Figure 4.4). This elbow is currently available for above elbow amputees and it utilizes EMG signals obtained from the biceps and triceps muscles to control the one degree of freedom (flexion/extension) of the elbow joint. In our prosthesis, we are not using EMG to control the position of the elbow, thus, only the mechanics of the Boston elbow are employed in our system. For the control, we are using the concepts described earlier in this thesis. Use of the Boston elbow's hardware, however, offers several advantages in our final prosthesis.

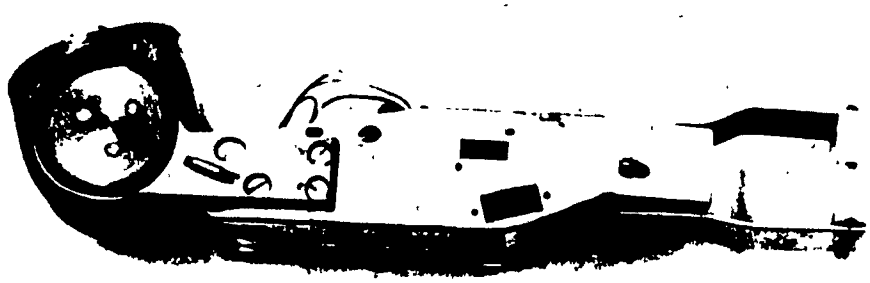

Fig. 4.4: Boston elbow.

The way the Boston Elbow is designed allows us to place the controlling electronics and the batteries inside the forearm. This will make the prosthesis self contained which is a very attractive feature from an aesthetic point of view.

Additional advantages in using the Boston Elbow are:

1) High speed - 145° of flexion in only one second.
2) Free swing for ease in walking – the elbow swings freely 30° or locks directly to motor.
3) Length adjustable to small or large amputees.
4) Reverse locking clutch - Arm locks unless motor is being driven.
5) High torque - 6.1 Newton-meters.
6) Can support 23 kg at 30 cm.
7) Overriding clutch protects against falls (this is very important when the elbow is being extended).
8) Routine maintenance required only once per year.

To adopt the Boston elbow to our system, all we need to do is to mount a potentiometer in its elbow motor housing. Once this has been done, the Otto Bock wrist and hand units will be attached to the elbow. After placing the controlling electronics in the forearm, our prosthesis will be ready for clinical application.

4.4 Goniometer:

One obstacle that remains to be solved is the goniometer to measure shoulder flexion and extension. Currently an experimental goniometer developed by the Rehabilitation Institute of Montreal that measures shoulder flexion/ extension and abduction is available. However as it can be seen from figure 4.5, this goniometer is attached to the back in a way that it is too cumbersome for a prosthesis which is worn every day. Also, it protrudes too much on the side.

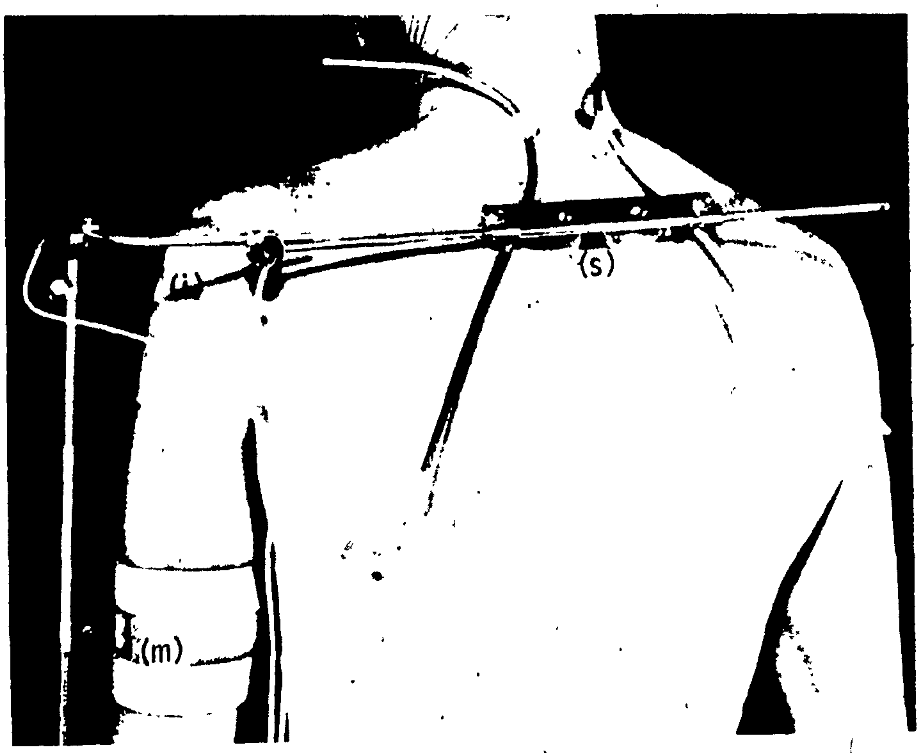

Fig. 4.5: RIM goniometer.

A new goniometer which can be placed under the arm and which is less cumbersome to wear has to be designed. An additional feature that must be included in the goniometer design is a braking mechanism to limit shoulder movement at any specified postion and time. This braking is necessary to make the servo "unbeatable".

Depending on the output capabilities of the sytem, the microprocessor controller will output a braking command. This command will remain until the system has reached the desired ouput demanded by the input, or if the shoulder movement is attempted in the opposite direction to that which caused the braking. This implies that the braking mechanism has to be bidirectional with fast application and release times.

At the present time such a goniometer is not available. Hence, for experimental purposes we have simulated the shoulder position input via a potentiometer and the braking commands are outputted to LED's telling which direction the braking should be applied.

5. ELECTRONIC HARDWARE

The controlling electronics for our above elbow prosthesis is based on an 8748 microcomputer manufactured by Intel Corporation [11]. The Intel 8748 is an 8 bit microcomputer system with 1K ROM (Read only memory) and 64 bytes RAM (Random access memory) integrated on a 40 pin chip. It has a fairly large set of instructions with an execution speed of 2.5 or 5 microseconds per instruction when running with a 6 MHz clock. Through its two I/O ports and its BUS port, the Intel 8748 provides sufficient I/O capabilities for most small microprocessor applications. It has an internal counter/timer circuitry which can either be used as a timer or an event counter independently.

Even though more powerful microprocessor systems with larger memory and I/O capabilities are available, and it is possible to expand the Intel 8748 system, we have found the single chip 8748 sufficient for our prosthesis application. We have chosen to use the 8748 system primarily because of its small size. It requires no other supporting hardware besides a crystal to supply the clock pulses. This compactness is very critical in our application since we wish to be able to mount all the controlling electronics in the elbow housing of the Boston Elbow, eliminating the need for an external control unit.

The ability to have a prosthesis with integrated control unit and power supplies will make the prosthesis more aesthetic and, at the same time, more practical to wear (since the amputee will not have the burden of attaching any external devices to some other part of his body or carrying them in his pockets).

Furthermore, the Intel 8748 has a CMOS version available (NEC μPD80C48) which has a very low power dissipation. This will contribute favourably towards having a prosthesis with integral power supplies, which has a sufficiently long daily life to make it practical.

The rest of the controlling electronics consist of an analog multiplexer (MPX), an analog to digital converter (ADC) and the driving electronics for the actuators.

A block diagram of the electronic hardware for the control of the elbow and wrist unit is shown on figure 5.1. The electronics for the control of the hand prehension developed by the Rehabilitation Institute of Montreal will not be included in this thesis since it is not part of the work described here.

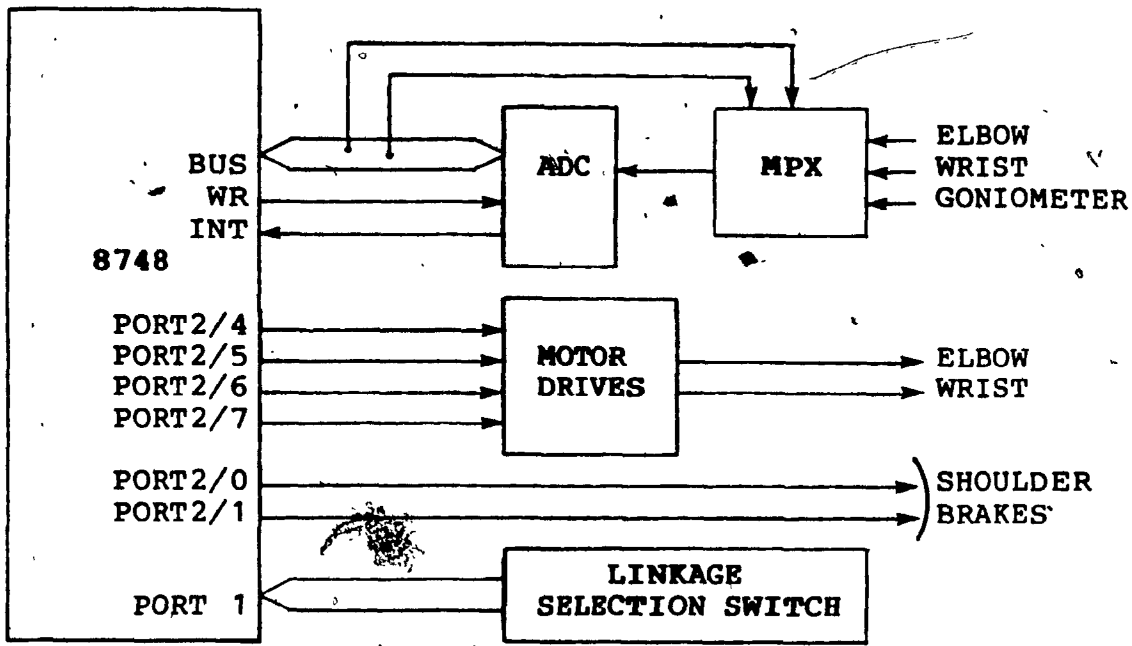

Fig. 5.1: Electronic Hardware Block Diagram.

Two linear potentiometers mounted on the elbow and the wrist unit provide the position information on the respective prosthesis joints to the microcomputer. A third potentiometer from the goniometer provides the position information on the amputee's shoulder joint. Those three signals provided by the potentiometers are fed into an analog multiplexer which is under software control. The analog signal selected by the software is converted into an 8-bit digital value by the ADC which also is under software control.

The BUS port of the microcomputer is used to send the commands to the analog multiplexer and also to access the digital data provided by the ADC.

As an input port, the bits D_0-D_7 of the BUS represent the selected analog signal as an 8 bit digital word. As an output port, the bits of the BUS are used as follows:

D_7	D_6	D_5	D_4	D_3	D_2	D_1	D_0

D_7–D_2: not used

D_1 D_0:

```
0 0   select MPX 0 (elbow position)
0 1   select MPX 1 (goniometer)
1 0   select MPX 2 (wrist position)
1 1   select MPX 3 (not used)
```

Fig 5.2: Bits of BUS as an output port.

A fourth input to the microcomputer comes from an 8 position BCD switch. This switch is under patient control and will allow him to select the task he wishes to peform with his prosthesis. The switch is connected to PORT 1 of the 8748. The bits of this port are used as in figure 5.3.

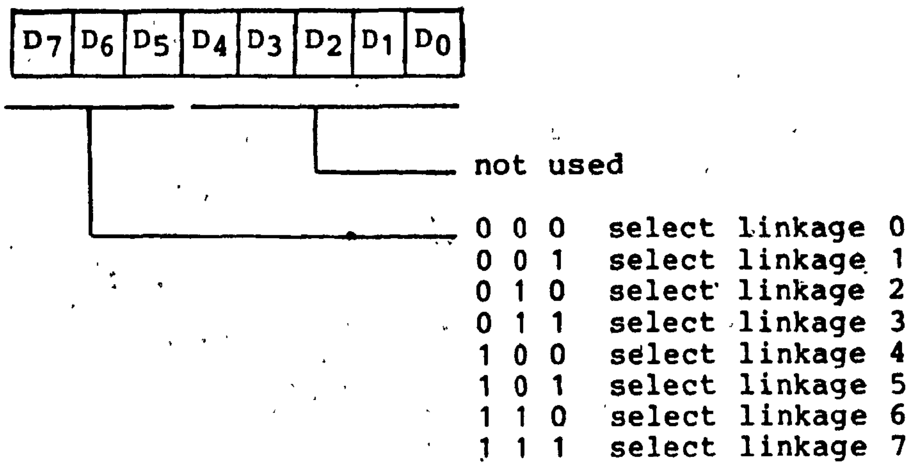

Fig 5.3: Input bits of PORT 1

The three most significant bits (bit 7 through 5) serve the software as an offset to the one of eight preprogrammed linkage look-up tables stored in the highest 256 byte bank of the microcomputer ROM. The other five bits of this port are not currently being used. They may be used at a later stage where there is a requirement to accommodate and access more than the present limit of 8 linkage tables.

PORT 2 of the Intel 8748 is used as an output port to activate the elbow and wrist motors and the bidirectional shoulder brakes for the accomplishment of the unbeatable servo feature. The bits of PORT 2 are used as shown in figure 5.4.

D7	D6	D5	D4	D3	D2	D1	D0

not used

shoulder brakes
0 0 no brake
0 1 flex brake
1 0 extend brake
1 1 not used

wrist motor
0 0 no drive
0 1 +ve drive (wrist pronate)
1 0 not used
1 1 -ve drive (wrist supinate)

elbow motor
0 0 no drive
0 1 +ve drive (elbow flex)
1 0 not used
1 1 -ve drive (elbow extend)

Fig. 5.4: Output bits of PORT 2.

The shoulder brakes, even though not yet available, will be made bidirectional. This means that the shoulder can be restrained from moving further in the direction that initiated the braking, whilst allowing movement in the opposite direction. This will allow the patient to move his stump back from the original position which will eliminate the possibility of jamming of the prosthesis against large loads above the handling capabilities of the motors. The direction at which the brakes should be applied, if at all, are determined by bit 0 and bit 1 combinations of the byte outputted to PORT 2.

6. THE SOFTWARE

The controlling software resides in the lower two 256 byte banks of the 1K ROM on board the Intel 8748. The highest bank is designated for the storage of the linkage look-up tables. The memory map is shown in figure 6.1.

The primary functions of the controlling software are:

(i) Monitoring the shoulder position of the amputee and determination of the desired elbow and wrist positions using the look-up table for the selected task.

(ii) Monitoring the current position of the prosthesis elbow and wrist joints.

(iii) Computing the optimum drive to achieve the desired position of the prosthesis and activating the motors to achieve this position.

(iv) Monitoring the current capabilities of the prosthesis and activating the brakes if necessary.

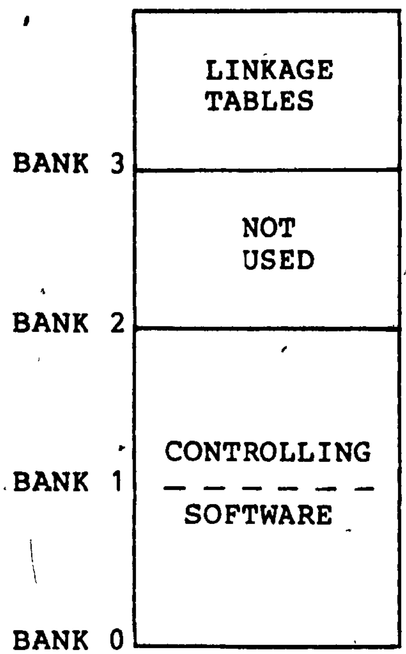

Fig. 6.1: Microcomputer ROM usage.

The following simplified flowchart (Fig 6.2) describes the basic steps involved in the control of the elbow joint (See Appendix A for a detailed flowchart of the software). The same algorithm is used for the control of both joints with the exception that the wrist is not monitored for unbeatable servo. This decision was based on the assumption that most normal movements will require the simultaneous activation of both joints and, for most movements, the wrist will be much less likely to fall short of satisfying the demands on the prosthesis than will the elbow.

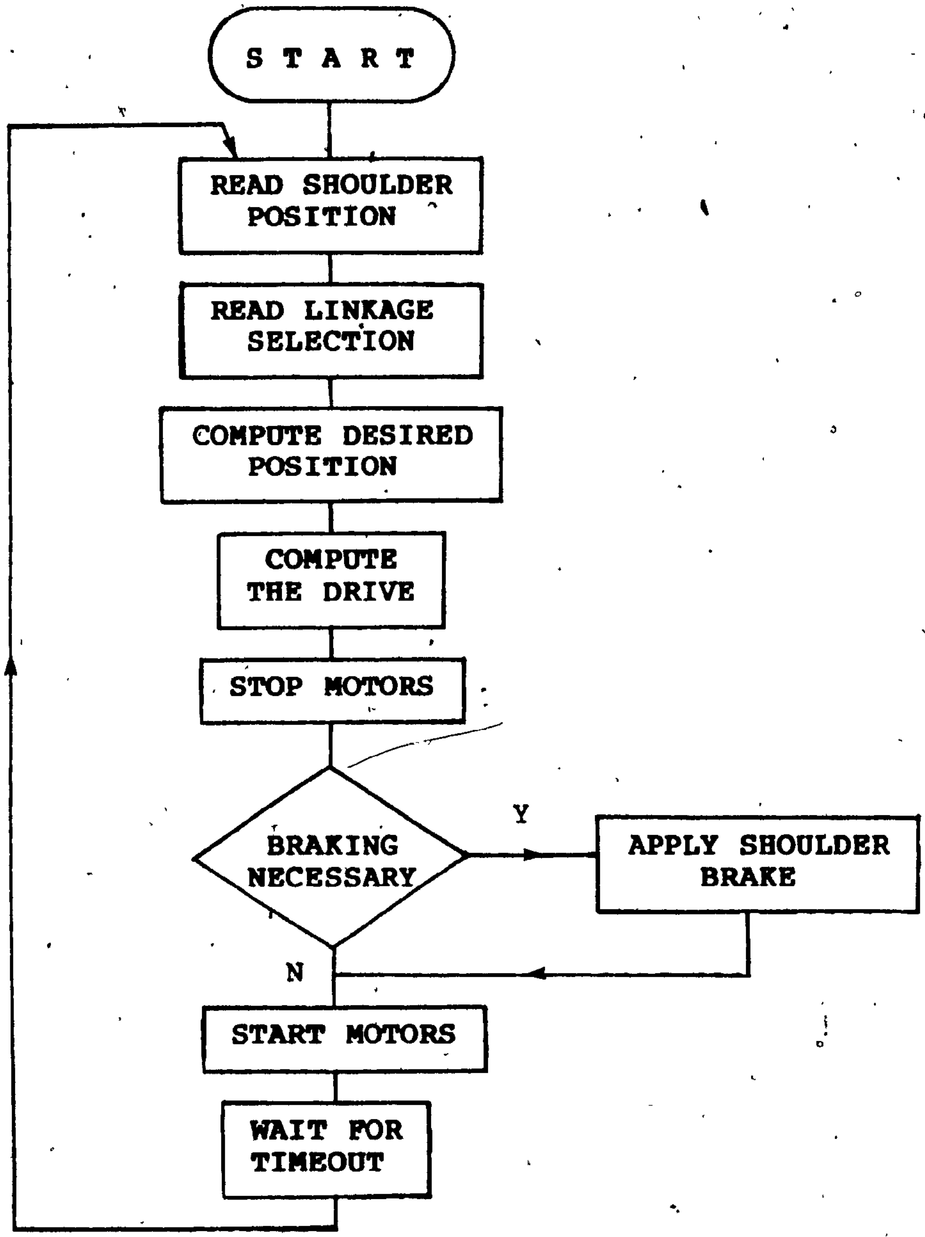

Fig. 6.2: Simplified flowchart for the control of the elbow joint.

The primary input to the microprocessor comes from the goniometer. The goniometer signal provides the communication from the amputee to the prosthesis. The microprocessor, having obtained the shoulder position information via this goniometer signal, computes the desired wrist and elbow positions from a look-up table. The look-up table to be used for the computation of the respective desired positions corresponding to the shoulder position, is determined by an external switch as mentioned earlier in chapter 3.

The microprocessor, having computed the desired wrist and elbow positions, proceeds to compute the necessary drives for the respective joints to reach their desired positions. To compute the necessary drive, the previous positions of the joints and their velocity are taken into consideration. The computation of the drive will be described in more detail later.

Having computed the necessary drives for both controlled joints, a signal, which is pulse-width modulated (with 23 msec period), is applied to the motors. The width of the pulses applied to each motor are determined by the computed drive for each joint, with a maximum pulse width of 20 msec.

If the computed drive is above the limits of the prosthesis, brakes are applied to the shoulder to restrict the amputee from further demands that can worsen the situation. The simultaneous activation of both controlled joints are achieved by setting the appropriate bits of PORT 2 described in chapter 5. It should be noted that the duration of the pulses applied to either joint are independent of each other. This independence is achieved by first outputting to the port with the bits set for the joint that requires the longer pulse duration and, after the necessary delay, setting the other bits of the port. The hardware timer/counter is used for the establishment of the time frame.

6.1 Computation And Application Of The Drive:

The necessary drive to bring the prosthesis from its present position to its desired position is computed digitally using equation 1.

$$\text{Drive} = A(x_d - x_p) - B(x_{p-1} - x_p) \tag{1}$$

Where: A and B are positive constants.

x_d is the desired position of the particular joint

x_p is the present position

x_{p-1} is the previous position

The first component of equation 1 gives a measure of the distance the prosthesis joint has to travel to reach it's desired position. The second component acts as a first order approximation of the velocity of the joint for which the drive is to be computed. This velocity feedback modifies equation 1 by the amount the prosthesis joint has travelled since the last computation of the drive. It has the effect of proportionally reducing the drive if the joint is moving in the same direction as the intended motion, and if in the opposite direction, increasing over the basic drive figure. The constants A and B are determined experimentally to improve system dynamics by varying the effect of the second component on equation 1.

The present position (x_p) is directly read from the feedback potentiometer mounted on the joint for which the drive is to be computed.

The desired position (x_d) is calculated using a look-up table. The program reads the goniometer value representing the amputee's shoulder position and uses the look-up table entry corresponding to the selected linkage to compute the desired position for the elbow or wrist joint.

The same technique is used in the computation of the drive for both elbow and wrist joints. The only difference being that of the entries used in the look-up table for the elbow or the wrist, and the feedback potentiometers used for each joint.

Having computed the necessary drive, the problem remains to transform this value into an appropriate signal to drive the motors. One method is to increase the DC power available to the motor proportionally to the computed drive from equation 1. This method is known as analog control. A second method is to apply pulses with duty cycle proportional to the computed drive. Although it was reported by Ulen, and Wagner [26] that this second method, known as Pulse-Width Modulation is not more efficient than analog control in terms of power consumption, works of Lozac'h et.al. [15,16] have found some merits in application of Pulse-Width Modulation to prosthetic systems.

The electrical energy applied to a motor is transformed into mechanical energy only when it reaches a certain percentage of the nominal energy. This percentage is a function of the load on the system. Inability to reach this percentage of the nominal energy due to loading may be the cause of a current drain through the motors, although no displacement of the driven device is observed. In the case of a prosthesis, where the energy storage is limited, this unnecessary power consumption will undesirably shorten the daily usage time.

If the driving signal applied to the motors is an analog voltage level, the amputee will not have any means of knowing when the motors of a prosthesis are being activated if no motion can be observed. If the driving signal is pulse-width modulated, however, the discrete tremor generated by the pulses informs the patient that his prosthesis is operating. The acoustic level of the tremor is minimal and not noticeable to the ear. If the tremor continues and yet no motion of the prosthesis is observed, the amputee, being aware of the fact, will be able to abort the task he was trying to perform.

The effect of this vibration feedback will be similar to the force feedback on a natural arm. In the case of the natural arm, similar to a prosthesis system, some tasks have to be aborted if the arm is not capable of handling the opposing force.

Having decided to use pulse-width modulation to drive the motors of our prosthesis, further investigations had to be made to find the optimum way of modulating the pulses relative to the computed drive. The translation of the calculated drive to the actual motor drive (voltage) would seem to require a simple relation of the form of figure 6.3.

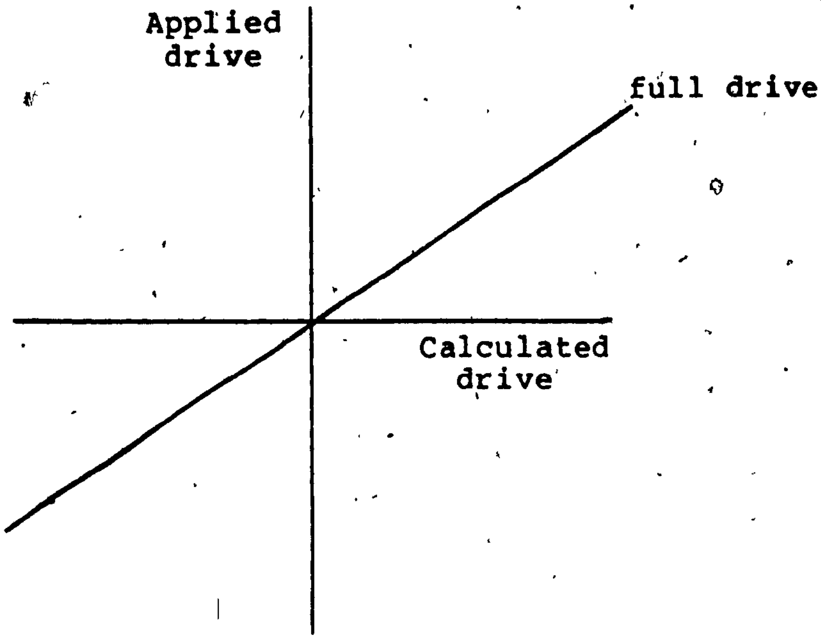

Fig. 6.3: Calculated drive versus actual motor drive.

Hence, whether analog or pulse-width modulated, the voltage level or duty cycle of the pulses respectively increase with increasing calculated drive. Further investigation with respect to the system under consideration shows that some modifications have to be made to the above rather simplified relationship as shown in figure 6.4.

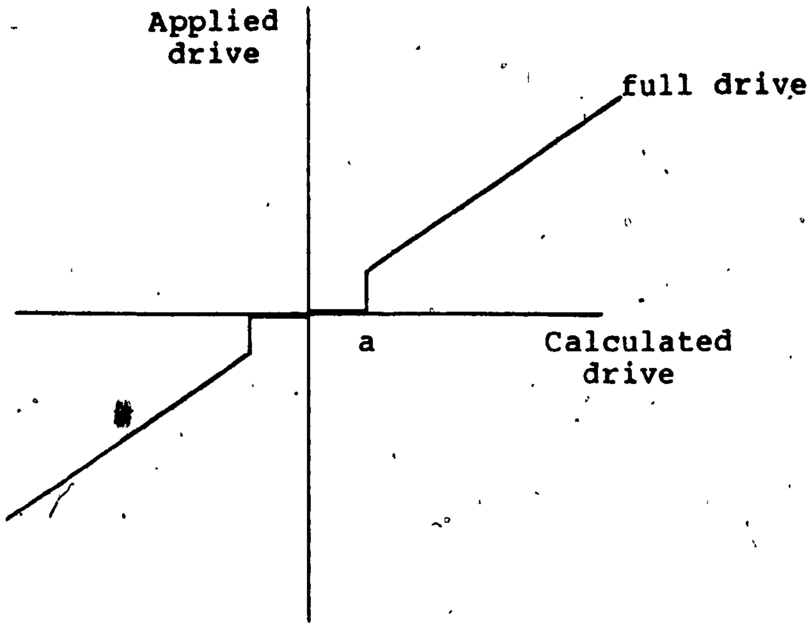

Fig. 6.4: Calculated drive versus actual motor drive.

Due to digital roundup errors and the possibility of slight unintentional movement of the patient's shoulder, the desired position - although the same as before - may be computed to be different. This will result in an undesirable jitter at the terminal device. The deadband introduced in the modified relationship of figure 6.4 eliminates slight variations in the computed drive, thus eliminating jitter. Secondly, it prevents current drain at the motors when no prosthesis motion is possible due to very low voltage levels or short pulses failing to reach the necessary percentage of the nominal energy to convert into mechanical energy (even when no load, besides the prosthesis itself, is present).

Further investigation of the drive relationship has shown the need for additional modification to improve the response of the prosthesis to the patient's commands. The final relationship used in our system is shown in figure 6.5.

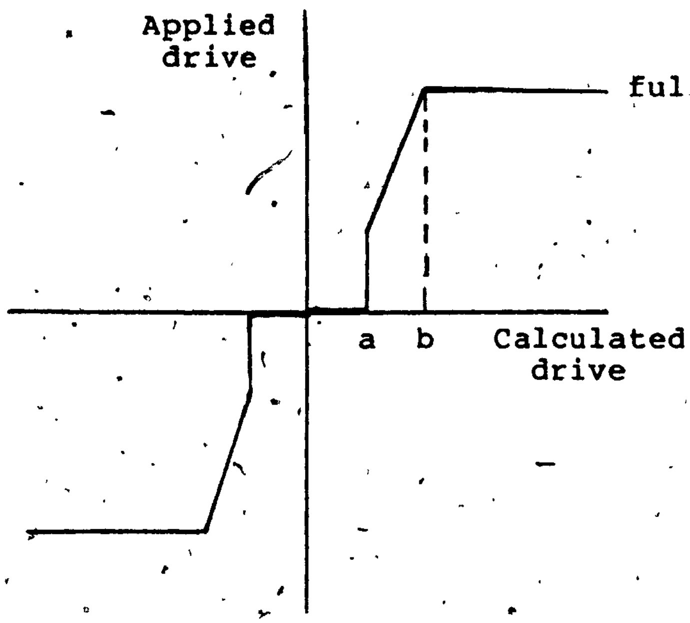

Fig. 6.5: Calculated drive versus actual motor drive.

The major reason for the modification is that the response of an artificial prosthesis with present technology is not as fast as a real arm, even if full drive is applied. If the proportional relationship of figure 6.4 is applied, it will lead to decreasing the drive as the present position closes on the desired position, reducing the response further. The reduction of speed as the desired position gets closer is a desirable feature, but experiments with our prosthesis have shown that, to ensure a faster response, a relationship as in figure 6.5 has to be used. This relationship ensures that full drive is applied when the prosthesis is beyond a certain distance (greater than b) from the desired position. At the same time it ensures that the applied drive decreases linearly as the prosthesis gets close to its destination (greater than a but less than b). This kind of approach is better than using an on/off kind of relationship as in figure 6.6, since the prosthesis does not stop abruptly as it falls into the deadband of the desired position. The limits a and b are determined experimentally by varying their values and monitoring the prosthesis response speed and it's smoothness as it gets close to the desired position.

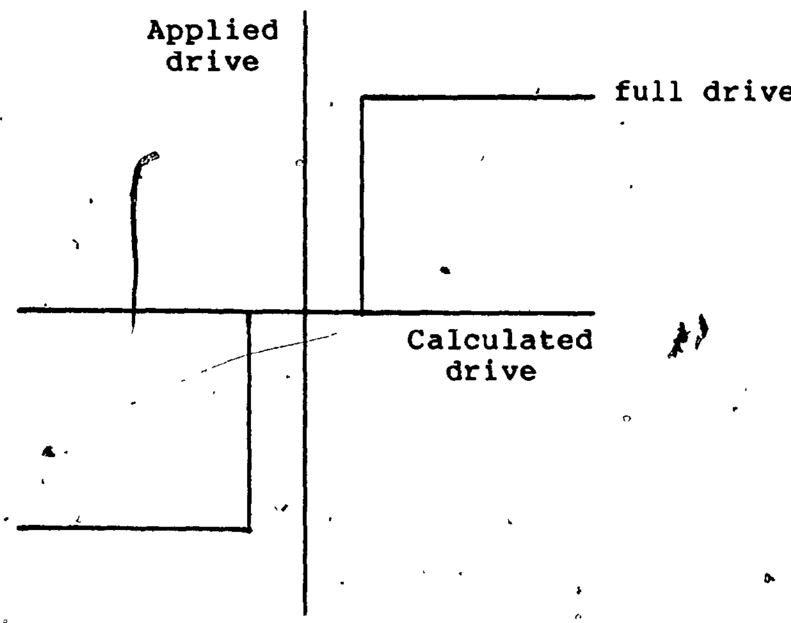

Fig. 6.6: Calculated drive versus actual motor drive.

Furthermore, with a relationship as shown in figure 6.6, it would not be possible to operate the prosthesis at different speeds. With the relationship of figure 6.5, slow movements of the shoulder will result in small changes in desired position calculations, and the computed drive will fall inside the region bounded by points a and b, enabling variable speed operation of the prosthesis.

6.2 Linkages:

The input/output relationship, that is, the desired wrist or elbow positions corresponding to the amputee's shoulder position for a given task, is determined by a linkage. The input is the shoulder position, and the outputs determined by the linkages are the wrist and elbow positions.

Regarding the input and output positions as angular displacements from a fixed reference, the relationship can simply be written as:

$$\theta = f_{1i}\ (\alpha)$$

Where θ is the wrist or elbow position and α is the shoulder position. f_{1i} is one of the possible linkages.

Graphically the relationship would have the form of one of the curves shown in figure 6.7.

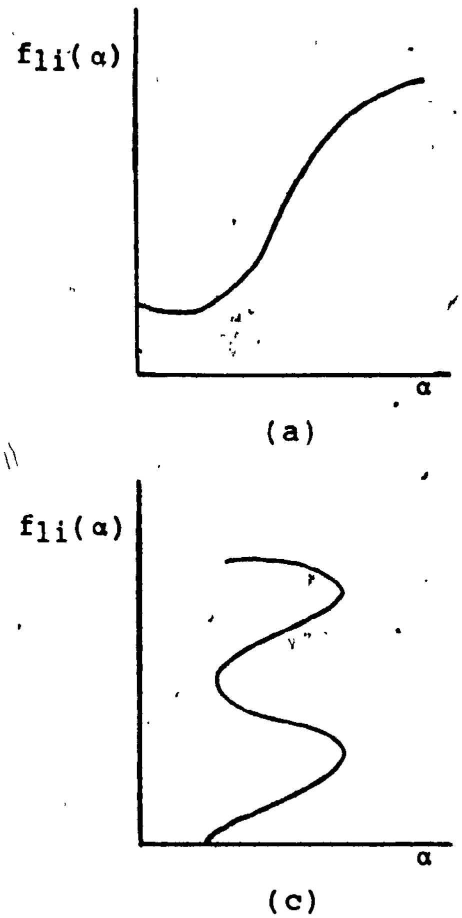

Fig. 6.7: Input/output curves for shoulder versus wrist or elbow for different linkages.

Although theoretically any of the above curves would be possible, in a position controlled prosthesis system, one cannot allow any linkages of the type (c). This is because a 1:1 relationship between the amputee's input via his shoulder position and the output wrist or elbow position must be established. A relationship of the type (c) would imply that there can be more than one elbow or wrist position output for a given shoulder position. This situation would be confusing for the amputee because it would be very difficult, if not impossible, to know which of the more than one possible outputs corresponds to his input at any given time.

In our prosthesis, a 1:1 relationship is maintained for all the linkages. Hence, a different output wrist or elbow position for the same input position is only possible under different linkages (i.e. when the amputee has selected a different linkage to perform another task).

Having determined the possible types of linkages, the second step is to implement those input-output relationships on the system under consideration.

Two approaches are possible to implement the linkages on a microprocessor controlled prosthesis. The first of those is to implement $f_{1i}(\alpha)$ analytically on the system for each task. Then for a selected task, the software will use the appropriate $f_{1i}(\alpha)$ to determine the output wrist and elbow positions θ, for each input α, as the amputeee moves his shoulder to perform a task. This scheme, even though very elegant, will require a lot of computation time. This overhead is especially un-acceptable in a prosthesis system because it will slow down the response of the prosthesis.

This slowed down response, along with the mechanical restraints on speed, will make the prosthesis too slow to be acceptable. Secondly, this scheme requires the prior knowledge of $f_{li}(\alpha)$ which may be very complex in many cases.

A second method is to use look-up tables to determine the output positions corresponding to the input positions. This method is much more suitable than the analytical approach to determine the input/output relationships in a microcomputer based system. The major advantage is in the considerable reduction in real time computations.

In the look-up table method, $f_{li}(\alpha)$ is solved for every possible input position α, and the values are stored in a table in microcomputer ROM. This is repeated for each linkage to be provided in the prosthesis. Thus for n linkages, n such tables have to be constructed and stored in memory. The tables can be constructed by either solving $f_{li}(\alpha)$ analytically, or, in cases where $f_{li}(\alpha)$ is difficult to determine, experimentally. In the experimental approach, every possible shoulder position and the corresponding elbow and wrist positions will have to be determined for each linkage and a table constructed.

One drawback of this table approach is that it has very large memory requirements. For our prosthesis, the shoulder will be able to move freely over 128°. Hence, if we were to store the output position for each degree of shoulder displacement, we would need 256 bytes (128 for the wrist and 128 for the elbow) per linkage. This would be impossible to store in our present memory capacity, hence, we would need external memory to accommodate the 2K (256x8) bytes for our 8 possible linkages.

The addition of external memory and the hardware to access this memory would increase the space requirements of our controlling electronics and would increase the power consumption of the system.

A method to overcome the large memory requirements described earlier, is to piecewise linearize the function $f_{1i}(\alpha)$. We have done this piecewise linearization by splitting $f_{1i}(\alpha)$ into 8 equal regions as shown in figure 6.8.

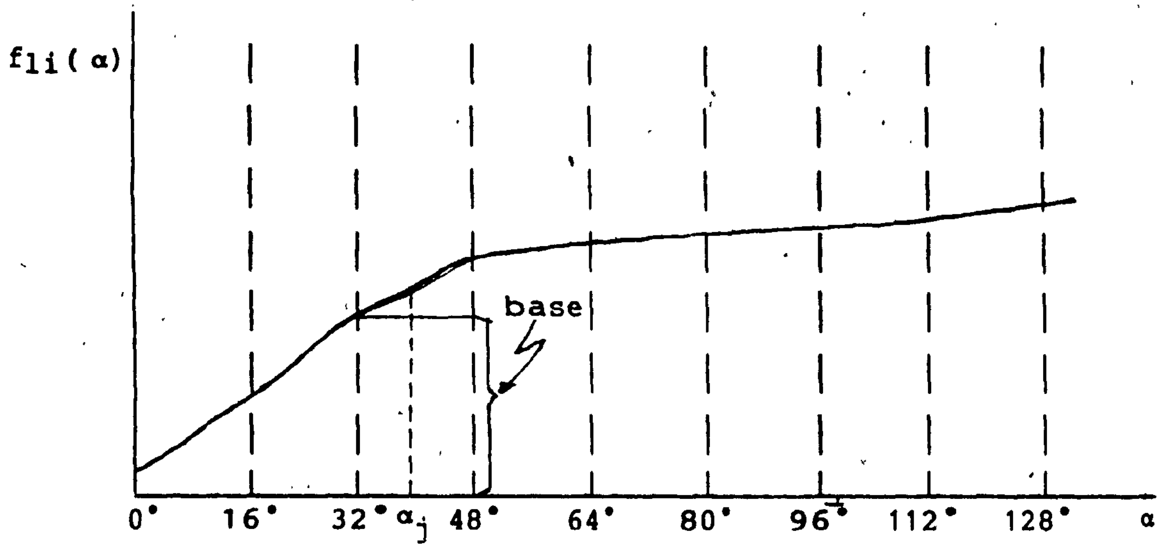

Fig. 6.8: Linearization process.

Each region covers 16° of the maximum possible 128° position range of the shoulder joint. Having split $f_{1i}(\alpha)$ into those 8 regions, we then have piecewise linearized the function in each region. The regions are small enough for our application not to cause unacceptable error. The choice of the number of the regions is made with the assumption that no task will require sharp slope changes within a 16° envelope of shoulder displacement.

The function $f_{li}(\alpha)$ is then approximated for each region by the equation:

$$f_{li}(\alpha_j) = M(\alpha_j - \alpha_{rk}) + C$$

Where M = slope

C = base

α_{rk} = α at the beginning of the k^{th} region

α_j = the input angle that falls in the k^{th} region

With this scheme, all that has to be stored is the slope and the base values for each region. The microprocessor can compute $f_{li}(\alpha_j)$ from this information very quickly and obtain the desired output wrist or elbow positions for every shoulder position input.

16 bytes of data storage are required to control one joint in a given linkage. Thus for two joints, 32 bytes are required. Hence, 8 linkages can be stored in 256 bytes which is what would have been necessary for each linkage if the curves were not piecewise linearized.

A detailed description of the construction of a linkage table is given in Appendix B.

7. CONCLUSION

A microprocessor - controlled prosthesis for above-elbow amputees has been described. The functions supported by the prosthesis are hand prehension, wrist rotation and elbow flexion/extension. Hand prehension is controlled by EMG signals from biceps and triceps muscles using the system developed at the Rehabilitation Institute of Montreal. Wrist rotation and elbow flexion/extension are controlled, in a coordinated manner, using a modified form of e.p.p. (extended physiological proprioception) originally introduced by D.C. Simpson. The shoulder flexion/extension has been selected as the controlling input. The relationship between input and output is determined by a set of selectable linkages stored in microcomputer memory.

In designing our prosthesis, the most important factor we have considered is amputee acceptability. No matter how elegantly designed, a prosthesis system is to be considered useless unless it satisfies the amputee's needs. A survey through the related literature shows that the major reasons for the rejection of powered prosthesis systems by the majority of amputees are: lack of functionality, uncosmetic appearance, unreliability, excessive weight and difficulties in wearing and operating the prosthesis.

Cosmetic appearance appears to be one of the most important factors affecting amputee acceptability. Although less functional than a hook, a simple pinch prehension hand is more likely to find amputee acceptance because it resembles a normal hand and thus will help to restore the normal appearance of the person.

Similarly, a prosthesis system which is bulky and needs the attachment of external components, such as battery packs, harnesses etc. is less likely to be accepted by amputees. The prosthesis described in this thesis is expected to satisfy the aesthetic needs of the amputee because it is self contained, with integral power supplies, and has the appearance of a natural arm.

However, as mentioned earlier, aesthetics are not all that is required in a prosthesis. Functionality, ease of operation, and reliability are also major factors affecting amputee acceptability. Our prosthesis control strategy incorporates some features that contribute to improve those aspects.

One problem with existing prosthesis systems is that the complexity of operation increases with increasing functionality. By using preprogrammed linkages in our prosthesis, we have been able to reduce this complexity. The ability to simultaneously control the elbow and wrist joints of the prosthesis with one body input (the shoulder position) will reduce the demands on amputee concentration. The use of the shoulder joint as the input will be a help rather than a hindrance, since most manipulative tasks do require the coordinated motion of shoulder, elbow and wrist joints.

Earlier, it was stated that a prosthesis system with fixed linkages would limit the amputee's ability to perform more than a few tasks. The selectability of the linkages in our prosthesis will allow the amputee to choose the appropriate linkage for the intended task, thus overcoming this limitation. Currently, we are able to incorporate eight linkages and these are not necessarily limited to the same number of tasks. Once the amputee becomes well accustomed to the different linkages available in his prosthesis, he may use a given linkage to perform several different tasks that require similar linkages.

Another important factor in a prosthesis, is amputee confidence in its performance. Proprioceptive feedback, provided by employing extended physiological proprioception in our prosthesis, will help to develop this amputee confidence.

Having learned the linkages available to him, the amputee will be able to associate his shoulder joint positions with the prosthesis joint positions. This is different from the EMG controlled prosthesis where the amputee is not aware of the electrical activity in his muscles and thus has to rely on visual observation to ensure that the prosthesis is behaving according to his demands. The proprioceptive feedback in our prosthesis will release the amputee from this burden and the operation of the prosthesis will become more subconscious. This is extremely important since a prosthesis system, unlike other machines operated by men, should not require high levels of concentration by the amputee. A person wearing a prosthesis should be able to be involved in other activities, such as thinking, talking or looking somewhere else while performing a task.

Ideally an arm prosthesis should replace all the functionality of the lost limb. However, technological limitations, especially mechanical constraints, restrict us from developing an artificial limb that will match the functionality of the natural limb. Thus a prosthesis, at its best, is still a poor replacement for the natural limb. Even so, it provides the amputee with a lot of functionality.

The prosthesis described in this thesis will be most effective for amputees who have the use of one hand. Naturally, an amputee who has the use of one arm will use this arm most of the time for all the tasks that he can perform with one hand. However, for tasks that involve the use of two hands, the artificial arm will be used as a helping hand, while the normal arm is performing the task.

People who are involved in jobs that are intellectually oriented will not need this helping hand most of the time. For them, a prosthesis perhaps won't be more than a cosmetic device to restore their physical appearance. For people who are involved in jobs that are manually oriented, however, the loss of one arm may mean that they will not be able to continue their profession. To learn a new profession that does not require the use of hands, will be very difficult in most cases. A prosthesis that can restore at least some of the functions of the lost arm will be indespensible for this class of amputees.

A prosthesis is most essential for bilateral amputees to restore some of the functions of the normal arm and for unilateral amputees that are involved, or wish to get involved, in manually oriented professions. Our prosthesis has advantages over the existing prosthesis in terms of serving this latter class of amputees. The use of look-up tables in determining prosthesis input/output relationships, allows for the easy modification of the functions of our prosthesis according to individual amputee's needs. Thus, we are able to provide the amputee with custom designed prosthesis functions. This flexibility is very important because different work environments require different manipulative abilities. An amputee working in a machine shop may need, as one of his selections, a linkage to operate a drill press; whereas an amputee working in a carpenter shop may need a linkage to operate a bench saw.

The unique feature of the above-elbow prosthesis described in this thesis is that many different linkages can be pre-programmed in microproessor memory. An additional feature is that the system dynamics are controlled by the microprocessor thus allowing variable speed operation of the artificial limb. It must be noted that the prosthesis is under total control of the amputee, hence there is no robotization.

The controlling concepts used in this current prosthesis can be extended to include the shoulder joint as one of the controlled functions of the prosthesis for shoulder disarticulation amputees. The movement of the clavicles will then have to be used as the controlling input, and the linkage tables have to be extended to incorporate the input/output relationships for the shoulder joint.

Further possible expansions are the incorporation of passive wrist bending and elbow rotation mechanisms. The necessity for such features has to be determined at a later stage.

8. REFERENCES

(1) Carlson, L.E. and Primmer, K.R., Extended Physiological Proprioception for Electric Prosthesis. In: Proc. of the Sixth Internation Symposium on External Control of Human Extremities. Suppl. Yugoslav Committee for Electronics and Automation. Belgrade, pp. 201-211, 1978.

(2) Carlson, L.E., Multi-Mode Control of an Above-Elbow Prosthesis, D. Eng. Dissertation, Technical Report 61, Biomechanics Laboratory, University of California, Berkeley, Sept. 1971.

(3) Carlson, L.E., An Above-Elbow Prosthesis with Optimized Coordination. Proc. of 5th Int. Symp. on External Control of Human Extremities, ETAN, Yugoslavia, Dubrovnik 1975.

(4) Childress, D.S., Powered Limb Prostheses: Their Clinical Significance. IEEE Transactions on Biomedical Engineering, Vol. BME-20 No.3, May 1973

(5) Freedy, A., and Lyman, J.H., A microcomputer based myoelectric pattern controlled arm prosthesis, Proceedings 5th Sumposium on External Control of Human Extremities, Dubrovnic, Yugoslavia, 1975.

(6) Gibbons, D.T., O'Riain, M.D., and Reuben, K.A., A Microprocessor Controlled Arm Prosthesis, Proceedings of the 8th annual Northeast Bio-Enginnering Conference, Boston, March 1980, pp. 106-109.

(7) Graupe, D., Magnussen, J., and Beex, A.A., A microprocessor System for Multifunctional Control of Upper-Limb Prostheses via Myoelectric Signal identification. IEEE Transactions on Automatic Control, vol. AC-23, 538-544, 1978.

(8) Graupe, D. and Cline, W.K., Functional Seperation of EMG Signals Via ARMA Identification Methods for Prosthesis Control Purposes, IEEE Trans. Systems, Man and Cybernetics, vol. SMC-5, pp.252-259,march 1975.

(9) Graupe, D., Control of Upper Limb prosthesis in Several Degrees of Freedom. Bulletin of Prosthetics Research, pp. 226-236, Fall, 1975.

(10) Hogan, N., A review of the methods of processing EMG for use as a proportional control signal, Biomedical Engineering, March 1976, 81-86.

(11) INTEL, MCS-48 Family of Single Chip Microcomputers. User's Manual, April 1979.

(12) Jacobsen, S.C., and Mann, R.W., Control Systems for Artificial Arms, IEEE Conf. on Systems, Man and Cybernetics, November 1973.

(13) Lambert, T.H., and Hall, M.J., Design And Control of Powered Artificial Arms. Proc. Instn. Mech. Engrs., Vol 183, pt 3J, pp. 1-6, 1968-69.

(14) Lawrence, P., Computer Design and Simulation of a Myoelectric System for Controlling a Multifunctional Prosthetic Hand. Proceedings of Symposium on Pattern Recognition and Mathematical Image Processing, Gotenborg, Sweden, August 4, 1972.

(15) Lozach, Y., Sherman, D.E., and Gingras, G. A New Versatile Myoelectric Control Unit for Clinical Application. Proc. of the 4th Int. Symposium on External Control of Human Extremities. Dubrovnic, 1972.

(16) Lozac'h, Y., Lippay, L.A., Sherman, D.E., and Gingras, G., Helping Hands, Electronics pp. 124-131, Aug. 7, 1967.

(17) Lyman, J.H., Freedy, A., Prior, R., and Solomonow, M., Studies Toward a Practical Computer-Aided Arm Prosthesis System. Bulletin of Prosthetics Research, pp. 213-225, Fall 1974.

(18) Neff, G., Prosthetic principles in bilateral shoulder disarticulation or bilateral amelia. Prosthetics and Orthotics International, 1978, 2, pp. 143-147.

(19) O'Riain, M.D., Gibbons, D.T., and Reuben K.A., Extended Physiological Proprioception in a Microprocessor Controlled Prosthesis. Proc. of Int. Conference on Rehabilitation Engineering. Toronto, Canada, 1980, pp. 47-50.

(20) Sheridan, T.B., and Mann, R.W. Design of Control Devices for People with Severe Motor Impairment. Human Factors, 20 (3) pp. 321-338, 1978.

(21) Simpson, D.C., The Choice of Control System for Multi-movement Prosthesis: Extended Physiological Proprioception (epp). In: The control of upper-extremity prosthesis and orthoses, Eds: Herberts, P.;Kadefors, R., Magnusson, R.; and Petersen, I.; Pub: Thomas, C.C., Springfield, Ill. pp. 146-150, 1974.

(22) Simpson, D.C., The control and supply of a multimovement externally powered upper limb prosthesis. Proc. 4th International Symposium on External Control of Human Extremities. Dubrovnik, 1972.

(23) Simpson, D.C. Extended Physiological Proprioception (epp). Rep. Int. Symp. Control of Upper-Extremity Prostheses and Orthoses, Goteborg, 1971.

(24) Simpson, D.C., and Sunderland, G.D. The Development and Control of Powered Prosthesis for Children. Health Bull. Vol. 22 pp. 67-69, October 1964.

(25) Thompson, R.G., and Childress, D.S., Design of a Powered Arm for Shoulder Disarticulation Amputees. Bulletin of Prosthetics Research, p. 135, Fall 1977.

(26) Ulen, E., and Wagner, M., Myoelectric Control for Pneumatic Prosthesis, Annual Report, Rehabilitation Institute of Montreal, Vol. v, Oct. 1969

9. APPENDIX A

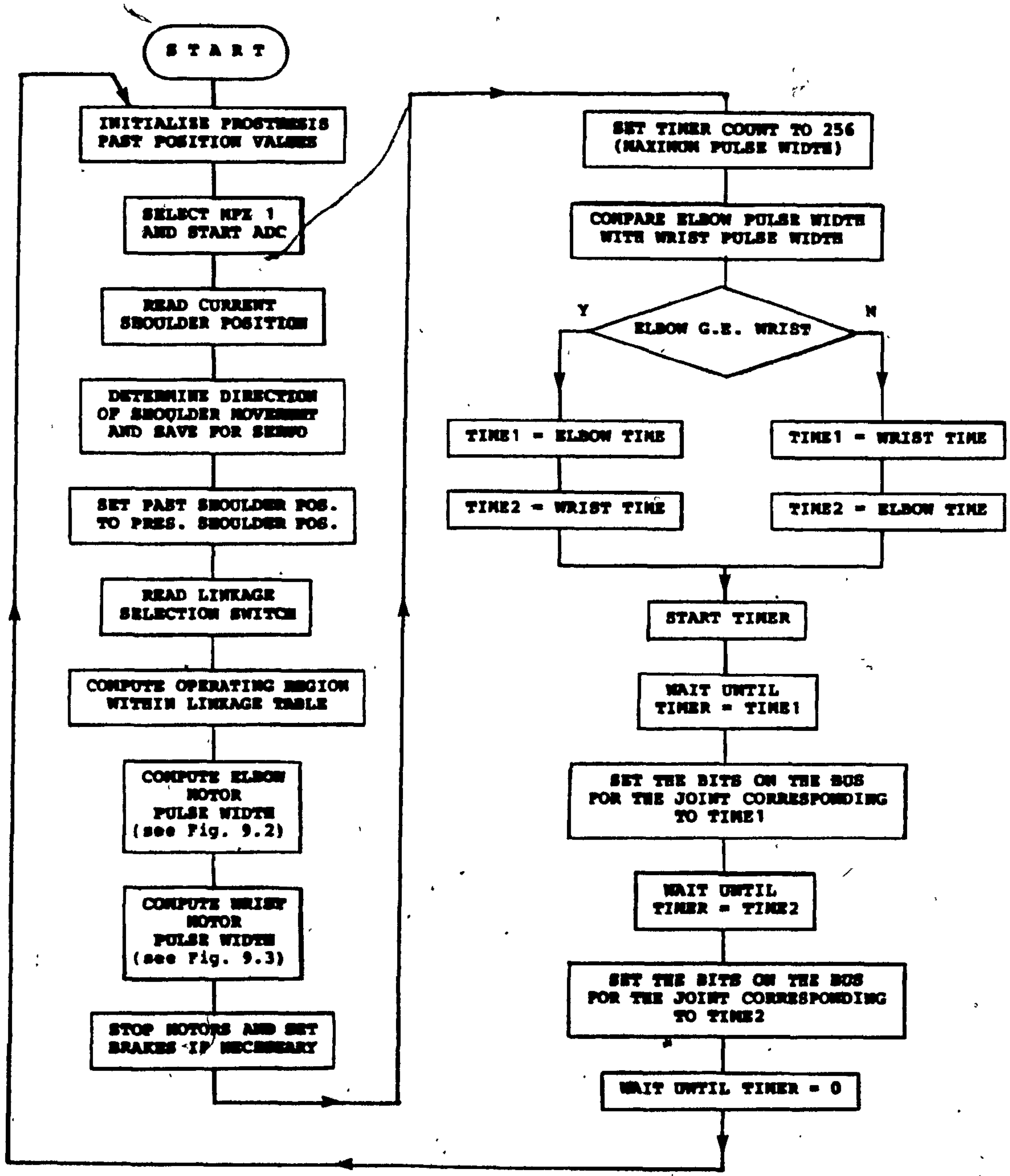

Fig. 9.1: Prosthesis control flowchart.

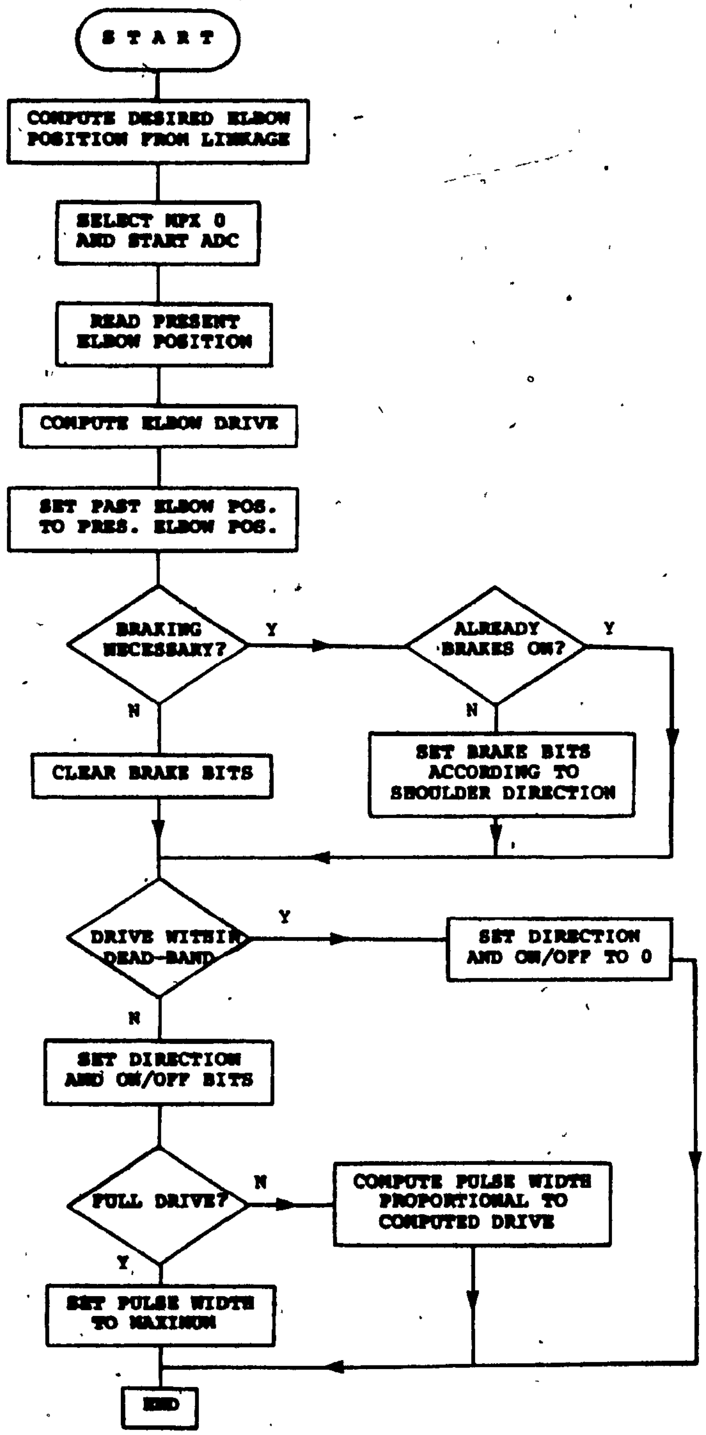

Fig. 9.2: Flowchart for the computation of the elbow pulse width.

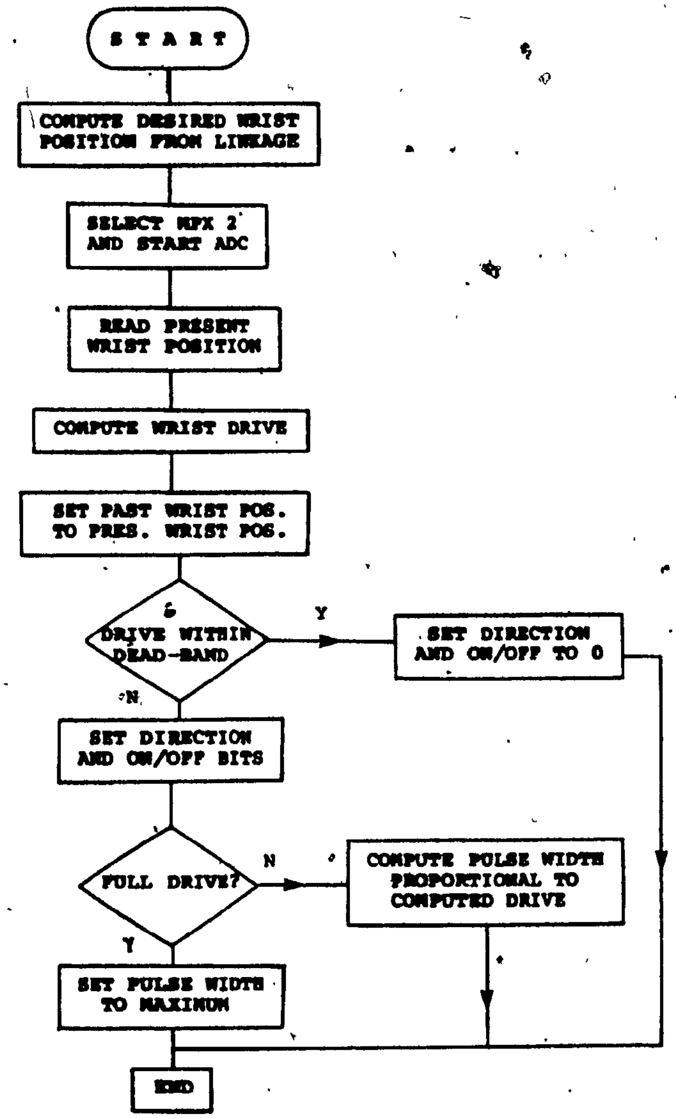

Fig. 9.3: Flowchart for the computation of the wrist pulse width.

10. APPENDIX B

In this appendix, as an example, the construction of the linkage table to operate a bench saw will be illustrated. In constructing this linkage table, it is assumed that the prosthesis will be used by an amputee who is approximately 6 feet tall. A graphical approach has been taken to establish the desired positions of the prosthesis elbow and wrist joints relative to the shoulder position.

The arm motions required to operate a bench saw can be graphically modelled as in figure 10.1, where:

a_1 = 12 in. (distance between shoulder and elbow joints)
a_2 = 13.5 in. (distance between elbow joint and the palm)
s = 16 in. (distance between shoulder joint and bench)

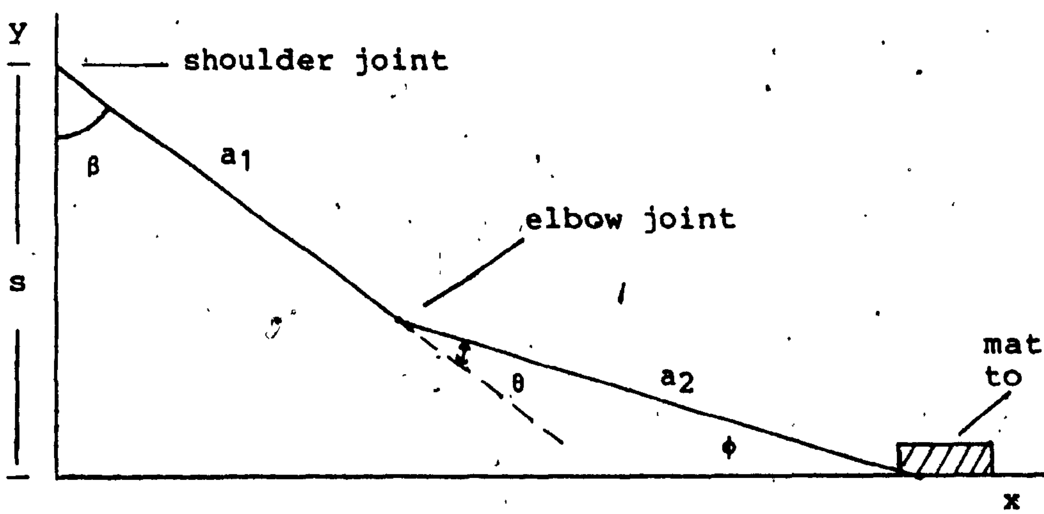

Fig. 10.1: Graphical representation of the natural arm movements to operate a bench saw.

To establish the linkage for the elbow motions, the relationship between β, (the angle between a_1 and the vertical axis), and θ, (the elbow angle measured as the displacement from its fully extended position), must be determined first.

From figure 10.1,

$$\theta = 90^\circ - \beta - \phi$$

and,

$$\phi = \sin^{-1} \frac{s - a_1 \cos(\beta)}{a_2}$$

hence,

$$\theta = 90^\circ - \beta - \sin^{-1} \frac{s - a_1 \cos(\beta)}{a_2} \tag{B1}$$

The goniometer displacement (α) is measured as shown in figure 10.2.

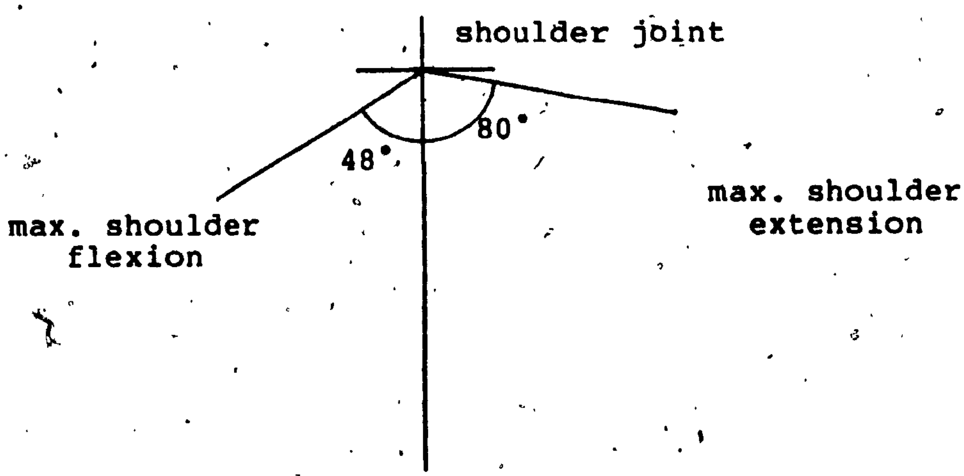

Fig. 10.2: Goniometer displacement versus shoulder angle.

The limitations on shoulder flexion and extension in figure 10.2 roughly define the envelope in which the mounting sockets are expected to allow the shoulder to move.

From figure 10.2, the goniometer angle (α) is related to the displacement from the vertical (β) in figure 10.1 by:

$$\alpha = \beta + 48^\circ \qquad \text{(B2)}$$

From equations B1 and B2,

$$\theta = 138^\circ - \alpha - \sin^{-1} \frac{s - a_1 \cos(\alpha - 48^\circ)}{a_2} \qquad \text{(B3)}.$$

Using equation B3, the values of θ for α ranging from 0 to 99 degrees have been tabulated in table 10.1. The reason for having 99 degrees as the upper limit is due to the values of a_1, a_2, and s, which at 99 degrees goniometer displacement would force the elbow to be fully extended. It must be noted that, for the sake of calculations, it is assumed that the person operating the bench saw with his prosthesis will maintain his vertical body position. In actual practice, if he bends towards the table, he may have to compensate for this change by adjusting the bench hight.

The values for α, and θ are then ploted on a graph in figure 10.3 for piecewise linearization. The base and slope values for each region obtained from the piecewise linearization are tabulated in table 10.2.

α	θ	α	θ	α	θ
0.00	101.81	34.00	85.17	67.00	50.84
1.00	101.62	35.00	84.39	68.00	49.52
2.00	101.51	36.00	83.60	69.00	48.19
3.00	101.18	37.00	82.78	70.00	46.84
4.00	100.92	38.00	81.95	71.00	45.47
5.00	100.65	39.00	81.11	72.00	44.09
6.00	100.36	40.00	80.25	73.00	42.69
7.00	100.05	41.00	79.37	74.00	41.28
8.00	99.72	42.00	78.47	75.00	39.85
9.00	99.37	43.00	77.54	76.00	38.40
10.00	99.01	44.00	76.63	77.00	36.94
11.00	98.62	45.00	75.69	78.00	35.46
12.00	98.22	46.00	74.73	79.00	33.96
13.00	97.80	47.00	73.76	80.00	32.45
14.00	97.37	48.00	72.76	81.00	30.92
15.00	96.92	49.00	71.76	82.00	29.37
16.00	96.45	50.00	70.73	83.00	27.80
17.00	95.96	51.00	69.69	84.00	26.22
18.00	95.46	52.00	68.63	85.00	24.62
19.00	94.94	53.00	67.56	86.00	23.01
20.00	94.40	54.00	66.47	87.00	22.37
21.00	93.85	55.00	65.37	88.00	19.72
22.00	93.28	56.00	64.25	89.00	18.05
23.00	92.69	57.00	63.11	90.00	16.36
24.00	92.09	58.00	61.95	91.00	14.65
25.00	91.47	59.00	60.88	92.00	12.92
26.00	90.84	60.00	59.60	93.00	11.18
27.00	90.19	61.00	58.39	94.00	9.41
28.00	89.52	62.00	57.17	95.00	7.62
29.00	88.84	63.00	56.94	96.00	5.81
30.00	88.04	64.00	54.69	97.00	3.99
31.00	87.42	65.00	53.42	98.00	2.03
32.00	86.69	66.00	52.14	99.00	0.26
33.00	85.94				

Table 10.1: Elbow angle versus goniometer angle for bench-saw operation linkage.

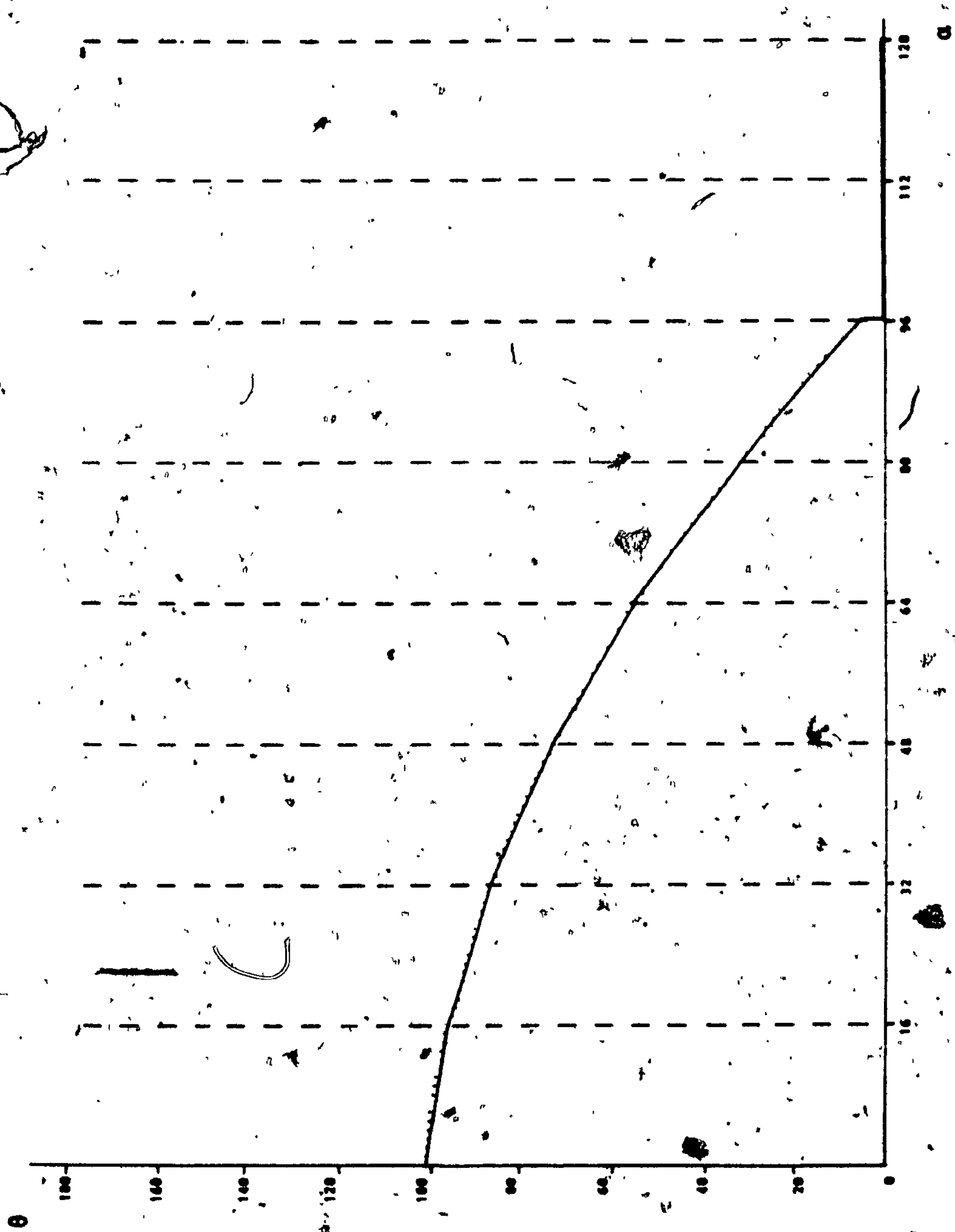

Figure 10.3 : Piecewise linearized bench-saw linkage.

REGION	BASE	SLOPE
1	101.81	-0.1675
2	96.45	-0.3050
3	86.69	-0.4353
4	72.76	-0.5647
5	54.69	-0.6950
6	32.45	-0.8325
7	5.81	0.0000
8	5.81	0.0000

Table 10.2: Base and slope values for each region of the elbow linkage for bench-saw operation.

The base and slope values of table 10.2 can now be converted into binary values interpretable by the microcomputer for the positioning of the elbow. The base values are approximated to the closest integer and are stored as an 8 bit binary word. The slope values are first divided by 2 and then stored as a signed 8 bit word as in figure 10.4. The reason for this division is to be able to make use of the full range of the 8 bit ADC via which the goniometer value is read. Each degree change of shoulder position is made to correspond to a change of a value of two at the ADC output for better resolution. Hence the slope's calculated from figure 10.3 have to be divided by two.

Using the above described approach, the final linkage table for the bench-saw operation is constructed in table 10.3. For this particular operation the wrist is stationary, hence, the derivation of the wrist values for the linkage table are greatly simplified. However, if this operation were to require the movement of the wrist, we would have to go through similar calculations as for the elbow to obtain the wrist entries of the linkage table.

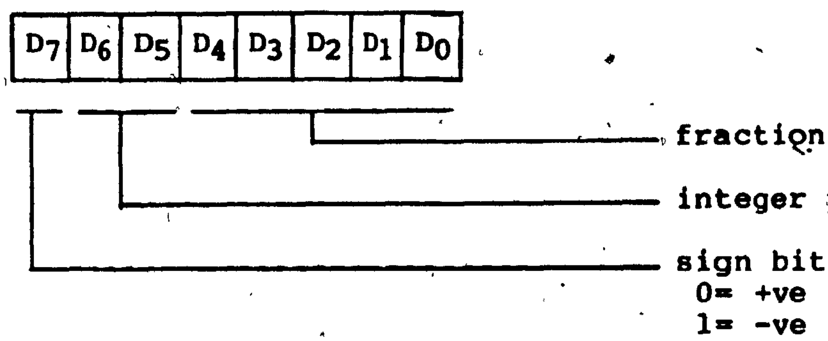

Fig. 10.4: Representation of the linkage table slope values as an 8-bit word.

Note: All negative numbers in the program are represented in 2's complement form. Numbers represented in the form shown in figure 10.4 are treated as being positive (i.e. MSB=0) for purposes of multiplication, and the result, if the number was negative, is converted into 2's complement. The multiplication routine used in the program is presented in Appendix C.

BINARY VALUE	HEX VALUE	MEANING OF ENTRY
10000101	85	Region 1 elbow slope
01100110	66	Region 1 elbow base
00000000	00	Region 1 wrist slope
01000000	40	Region 1 wrist base
10001010	8A	Region 2 elbow slope
01100000	60	Region 2 elbow base
00000000	00	Region 2 wrist slope
01000000	40	Region 2 wrist base
10001110	8E	Region 3 elbow slope
01010110	56	Region 3 elbow base
00000000	00	Region 3 wrist slope
01000000	40	Region 3 wrist base
10010010	92	Region 4 elbow slope
01001000	48	Region 4 elbow base
00000000	00	Region 4 wrist slope
01000000	40	Region 4 wrist base
10010110	96	Region 5 elbow slope
00111000	38	Region 5 elbow base
00000000	00	Region 5 wrist slope
01000000	40	Region 5 wrist base
10011011	9B	Region 6 elbow slope
00100000	20	Region 6 elbow base
00000000	00	Region 6 wrist slope
01000000	40	Region 6 wrist base
00000000	00	Region 7 elbow slope
00000110	06	Region 7 elbow base
00000000	00	Region 7 wrist slope
01000000	40	Region 7 wrist base
00000000	00	Region 8 elbow slope
00000110	06	Region 8 elbow base
00000000	00	Region 8 wrist slope
01000000	40	Region 8 wrist base

Table 10.3: Final linkage table for bench-saw operator.

11. APPENDIX C

```
;***** EIGHT BY EIGHT UNSIGNED MULTIPLICATION ROUTINE *****

;         INPUT:  R5=MULTIPLIER
;                 R4=MULTIPLICAND

;         OUTPUT: A =UPPER EIGHT BITS OF RESULT
;                 R5=LOWER EIGHT BITS OF RESULT


BMPY:     MOV     R6,H8          ; Set loop count to 8.
          CLR     A              ; Clear A.
          CLR     C              ; Clear carry bit.

BMP1:     RRC     A              ; Double shift right A & R5
          XCH     A,R5           ;  into carry.
          RRC     A
          XCH     A,R5

          JNC     BMP2           ; If carry not 1, don't add
                                 ;  multiplicand to A.
          ADD     A,R4

BMP2:     DJNZ    R6,BMP1        ; Decrement loop count and
                                 ;  loop if not zero.
          RRC     A              ; Final double right shift
          XCH     A,R5           ;  before exit.
          RRC     A
          XCH     A,R5
          RET
```

NOTE: Maximum time to execute this code is 0.205 msec. at 6 MHz clock rate, which is roughly 5% of the time it takes for the program to go through a cycle.